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# Long-distance walking effect on below-knee amputee compensatory gait patterns and implications on prosthetic designs and training

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**Abstract**

Trans-tibial amputees are advised to walk as much as able people to achieve healthy and independent life. However, they usually have difficulties in doing so. Previous researches only included data from a few steps when studying the gait of amputees. Walking over a long distance was rarely examined. The objective of this study was to investigate the changes in spatial-temporal, kinetic and kinematic gait parameters of trans-tibial amputees after long-distance walking. Six male unilateral trans-tibial amputees performed two sessions of 30-min walking on a level treadmill at their self-selected comfortable speed. Gait analysis was undertaken over-ground: (1) before walking, (2) after the 1st walking session and (3) after the 2nd walking session. After the long-distance walking, changes in spatial-temporal gait parameters were small and insignificant. However, the sound side ankle rocker progression and push-off were significantly reduced. This was due to the fatigue of the sound side plantar flexors and was compensated by the greater effort in the prosthetic side. The prosthetic side knee joint showed significantly increased flexion and moment during loading response to facilitate the anterior rotation of the prosthetic shank. The prosthetic side hip extensors also provided more power at terminal stance to facilitate propulsion. Endurance training of the sound side plantar flexors, and improvements in the prosthetic design to assist anterior rotation of the prosthetic shank should improve long-distance walking in trans-tibial amputees.

**Keywords**

effect, below-knee, long-distance, amputee, walking, compensatory, gait, patterns, implications, prosthetic, designs, training

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# LONG-DISTANCE WALKING EFFECTS ON TRANS-TIBIAL AMPUTEES COMPENSATORY GAIT PATTERNS AND IMPLICATIONS ON PROSTHETIC DESIGNS AND TRAINING

## Abstract

Trans-tibial amputees are advised to walk as much as able people to achieve healthy and independent life. However, they usually have difficulties in doing so. Previous researches only included data from a few steps when studying the gait of amputees. Walking over a long distance was rarely examined. The objective of this study was to investigate the changes in spatial-temporal, kinetic and kinematic gait parameters of trans-tibial amputees after long-distance walking. Six male unilateral trans-tibial amputees performed two sessions of 30-minutes walking on a level treadmill at their self-selected comfortable speed. Gait analysis was undertaken over-ground: 1) before walking, 2) after the 1st walking session and 3) after the 2nd walking session. After the long-distance walking, changes in spatial-temporal gait parameters were small and insignificant. However, the sound side ankle rocker progression and push-off were significantly reduced. This was due to the fatigue of the sound side plantar flexors and was compensated by the greater effort in the prosthetic side. The prosthetic side knee joint showed significantly increased flexion and moment during loading response to facilitate the anterior rotation of the prosthetic shank. The prosthetic side hip extensors also provided more power at terminal stance to facilitate propulsion. Endurance training of the sound side plantar flexors, and improvements in the prosthetic design to assist anterior rotation of the prosthetic shank should improve long-distance walking in trans-tibial amputees.

Keywords: Gait; Compensatory mechanism; Long-distance walking; Trans-tibial amputees.

## 1. Background

The ultimate goal of rehabilitation programs for trans-tibial amputees is to facilitate the best prosthetic function which allows them to undertake daily activities and achieve the highest possible activity level. However, amputees cannot be as physically active as able people [1], and are more susceptible to cardiovascular disease, including hypertension, hyperlipidemia and myocardial infarction [2]. The World Health Organization (WHO) states that people with disability are advised to undertake regular exercise and walking as much as able people [3]. Daily walking for over 30 minutes is recommended in order to attain a healthy life [4].

It has been suggested that in the modern community amputees are required to walk for longer distance and at higher velocity to achieve independent life [4]. However, another survey revealed that among 500 amputees interviewed 84% would walk outdoors for only 1-2 km at a time, and 60% of amputees could not tolerate walking for a long distance because they felt tired easily [5]. There is a need for better understanding of amputee gait over longer distance walking.

Previous gait analysis studies reported that trans-tibial amputees walk with shorter sound side step length due to the lack of prosthetic side ankle joint motion and greater ground reaction forces (GRF) at the sound side because of the sensation of pain at the prosthetic side [6-8]. As a result of perceived instability, they have decreased range of motion at the prosthetic side knee joint during walking [6-8]. This asymmetric gait pattern could lead to musculoskeletal disorders such as knee osteoarthritis at the sound side and scoliosis [13-15]. They walk more slowly but expend greater energy than able people when walking at self-selected comfortable speeds [9-12]. A variety of gait conditions have been investigated [16]: different gait initiation strategies, walking speeds, walking surfaces, alignments, types of amputation, prosthetic knee and feet designs as well as comparisons between fallers and non-fallers. All these studies provided valuable insights on how the amputee gait could be improved. Yet most of the published researches examined amputee gait for a few steps only. Gait changes when walking for longer distances have not been investigated.

The objective of this study was to investigate the changes in spatial-temporal, kinetic and kinematic gait parameters of trans-tibial amputees after performing long-distance walking. Such investigation could explain why amputees cannot walk for a long distance. It could provide useful information for developing strategies suitable for amputee training and for prosthetic design improvements to facilitate longer distance walking.

## 2. Methods

### *2.1. Subjects*

Six male subjects between 39 and 62 years of age, who had unilateral trans-tibial amputation performed at a minimum of 8 years prior to recruitment participated in the study. Their mean height was 170 cm (SD=3.4) and their mean body weight was 75 kg (SD=4.7). Immature residual limb with large volume change and walking tolerance of less than an hour were exclusion criteria. Detailed characteristics of the subjects are shown in Table 1. They were recruited from the Jockey Club Rehabilitation Engineering Clinic. Ethical approval was obtained from the Human Subject Ethic Sub-committee of The Hong Kong Polytechnic University and all subjects provided informed consent to participate.

*2.2. Testing protocol*

Subjects performed two consecutive sessions of 30-minutes walking at their self-selected comfortable speed on a level treadmill. The comfortable speed was achieved by instructing the amputees to adjust to the appropriate speed on the treadmill. Subjects were restricted from holding the handrail of the treadmill. For safety reasons, the test was monitored by a professional prosthetist and subjects could stop the test at any time. Collection of gait data when the subjects walked over-ground was undertaken (1) before the walking (baseline), (2) after the 1$^{st}$ 30-minutes walking and (3) after the 2$^{nd}$ 30-minutes walking. Trials were repeated until at least five samples of successful walking steps were collected, in which the whole foot completely landed within the force plate. All successful trials were used in the analysis. After treadmill walking, 2-3 minutes of time was given to the subjects to walk over-ground to get familiar with the changing environment. Each gait data collection session was completed within 5 minutes. Amputee subjects reported the change in lower-limb fatigue level and residual limb pain levels after one-hour walking using the visual analogue scale (VAS) marked on a 10-cm line [17].

*2.3. Apparatus*

Subjects used their own prostheses, with patellar-tendon-bearing sockets, throughout the experiment (Table 1). An eight-camera motion capture system (Vicon, Oxford Metrics) was used for three-dimensional motion analysis. Infra-red reflective markers were placed bilaterally on the dorsum of the foot, heel, lateral and medial malleoli, lateral and medial femoral condyles, greater trochanter, anterior superior iliac spine, posterior superior iliac spine, iliac crest and mid-shank and thigh. GRF was measured using two force plates (Advanced Mechanical Technology, Inc) embedded midway on a straight 8-m walkway. Kinematic and GRF data were collected at 200 and 1000 Hz respectively.

*2.4. Data analysis*

Gait data were analyzed using Visual3D (C-motion, inc). All kinematic and kinetic data were low-pass filtered using a $4^{th}$-order Butterworth filter with a cut-off frequency of 6 Hz. GRF in the anterior direction (propulsive force), in the posterior direction (braking force), in the medial-lateral direction (GRFy), in the vertical direction (GRFz), amplitudes of center of body mass (CoM) vertical movements, angle, moment and power at the ankle, knee and hip joints, walking speed, step lengths, stance time and cadence were measured. GRFz ratio, propulsive force ratio and braking force ratio were defined as the peaks of GRFz and GRFy in the prosthetic side divided by those of the sound side.

Kinetic data were normalized to the body mass. Local maxima and points of interest in kinetic and kinematic data were determined within the gait cycle for each successful walking trial. All spatial-temporal, kinetic and kinematic quantities were averaged in each test session, and then averaged across subjects. The Bonferroni-adjusted t-test for dependent sample was used to determine if there were significant differences between 1) the baseline and after the $1^{st}$ walking session, as well as 2) the baseline and after the $2^{nd}$ walking session. Significance level was set at $p < 0.05$.

**3. Results**

Before the two walking sessions, the peak sound side GRFz, propulsive force and braking force were significantly greater than that of the prosthetic side, with GRFz ratio of 0.87 ($p=0.016$), propulsive force ratio of 0.49 ($p=0.003$) and braking force ratio of 0.64 ($p=0.009$) respectively. Stance time of the sound side was significantly longer than the prosthetic side by 6% ($p=0.023$). Step length of the prosthetic side was significantly longer than the sound side by 13% ($p<0.001$).

All six amputees were able to complete the two sessions of 30-minute walking. After each walking session, changes in amplitudes of CoM and spatial-temporal gait parameters including walking speed, cadence, sound side and prosthetic side step lengths and stance time were very small and insignificant compared to the baseline (Table 2). The fatigue score in VAS for trans-tibial amputees was 7.9±0.8 after the 2$^{nd}$ 30-minutes walking, which was significantly higher ($p<0.001$) than the baseline of 3.0±0.8 (less fatigue). There were no significant changes in the residual limb pain score.

There were significant reductions in sound side peak propulsive force by 15% (1$^{st}$ session: $p=0.024$; 2$^{nd}$ session: $p=0.022$) and peak braking force by 9% (1$^{st}$ session: $p=0.017$; 2$^{nd}$ session: $p=0.005$) (Figure 1). The prosthetic side had significantly greater peak propulsive force by 21% ($p=0.02$) while it had significantly less peak braking force by 19% ($p=0.006$) after the 2$^{nd}$ 30-minutes walking (Figure 2). After the two walking sessions, the propulsive force ratio increased significantly by 39% ($p=0.003$).

After the 1$^{st}$ and 2$^{nd}$ 30-minutes walking, the prosthetic side knee joint had more flexion from 0% to 60% of the gait cycle. The first peak flexion angle was one-folded significantly higher ($p<0.025$) than the baseline after the 2$^{nd}$ 30-minutes walking.  No significant difference was found in sound side knee joint angles.

However, from 10% to 30% of the gait cycle the sound side knee joint moment decreased with the peak value significantly reduced by 26% ($p=0.025$) after the 2nd 30-minutes walking. During the same period of gait cycle, the prosthetic side knee joint moment increased significantly by 42% after the 1st 30-minutes walking ($p<0.025$) and 67% after the 2nd 30-minutes walking ($p=0.02$). At 60% of the gait cycle, the sound side knee joint moment decreased significantly by 15% ($p=0.005$).

After the 2nd 30-minutes walking, there was significantly less sound side dorsiflexion at around 30% of the gait cycle ($p<0.025$). There were also a trend toward 4% less dorsiflexion at around 50% of the gait cycle and 8% less plantar flexion at around 70% of the gait cycle. There was a reduction in sound side ankle joint moment from 25% to 62% of the gait cycle with the peak moment significantly reduced by 14% ($p=0.025$).

The sound side ankle power absorption at 20% of the gait cycle decreased significantly ($p<0.025$) after the 2nd 30-minutes walking. There was also significantly less sound side ankle power generation at 60% of the gait cycle (1st session: 9%, $p=0.014$; 2nd session: 12.2%, $p=0.013$). After the 2nd 30-minutes walking, the sound side hip joint generated significantly 45% more power ($p<0.025$) at 20% of the gait cycle, and the prosthetic side hip joint generated an average 10% more power at 60% of the gait cycle.

**4. Discussion**

The gait pattern of the six subjects before the long distance walking matched with previous gait analyses of trans-tibial amputees [6,7]. They relied more on their sound limbs by spending more time in stance phase and putting more forces against the ground compared to the prosthetic limb. After the two sessions of 30-minutes walking, muscle fatigue affecting the sound side would explain the reduced range of motion and moments of the sound side ankle and knee joints. However, the walking speed, step length, stance time, cadence and amplitude of CoM were not affected. Some compensation was made by putting more effort in using the prosthetic limb.

Gait analysis results suggest that the fatigue of the sound side plantar flexors led to some changes in gait pattern. There was a significant reduction in the sound side ankle power absorption during ankle rocker progression in mid-stance, and the plantar flexors generated significantly less power during terminal stance. These were signs of plantar flexors fatigue, as it was shown in previous studies on able people: following muscle fatigue they avoided shock absorption by reducing the eccentric contraction [18] and the joint power generation [19]. This is also in agreement with previous findings in able-bodies subjects, suggesting that muscles which act at the ankle are mostly affected by fatigue in long-distance walking [20]. Sound side plantar flexor fatigue led to reduced force at terminal stance and reduction of the joint power. This produced a significantly reduced propulsive force, ankle and knee moment at terminal stance, and a reduction in the average plantar flexion in early swing.

The fatigue of the sound side plantar flexors reduced the forward push-off. This was compensated by the significant increase in prosthetic side knee joint flexion at early stance phase. When the sound limb was at terminal stance, body weight was transferred onto the prosthetic limb [8,21]. As there was no active motion at the prosthetic ankle joint, the prosthetic foot rolled on the heel during loading response [25]. Prosthetic side knee joint flexion during loading response placed the prosthetic foot flat on the ground and allowed anterior rotation of the prosthetic shank [25]. The significantly higher prosthetic side knee flexion and moment at loading response following long-distance walking and fatigue would suggest that the prosthetic side knee joint played a more important role in bringing the body forward. It also led to a decrease in the prosthetic side braking force at loading response.

Plantar flexors also play an important role in providing stance stability [8,21-24]. As a result of the sound side plantar flexor fatigue, the sound side had reduced dorsiflexion angle at mid-stance. This placed the tibia in a more vertical position, which improved stability and allowed the line of action of the total GRF to be closer to the joint centres [6,18]. The sound side braking force, ankle and knee moments were therefore reduced, and the fatigue of the plantar flexor was relieved, to some extent. This may have some effects on gait as the dorsiflexion after loading response is one major source of forward body progression [8,21]. This was compensated by the significant increase in sound side hip power generation during loading response. It was interesting to note that significantly greater prosthetic side propulsive force was produced at the same time, while there was no active ankle motion. This increase was attributable to the prosthetic side hip joint generating an average 10% more power and the increased knee joint angle at terminal stance bringing the prosthetic shank in a more horizontal position. This was in line with previous studies comparing trans-tibial amputees with able-bodies subject which suggests that the increased hip extensors power of trans-tibial amputees was responsible for the forward progression of the trunk in the absence of active prosthetic ankle plantar-flexor activity [26,27]. The significantly increased prosthetic side propulsive force compensated for the reduction of sound side dorsiflexion after loading response.

The findings of this study have several clinical implications. 1) Attention is usually paid on the prosthetic side during rehabilitation. It should also be noted that endurance training for the sound side plantar flexors may improve long-distance walking in trans-tibial amputees. 2) Significantly increased prosthetic side knee joint flexion was noted during loading response after the long distance walking. Prosthetic designs which assist the anterior rotation of the prosthetic shank may be helpful. 3) Gait analysis is usually used to compare new prosthetic components [16]. Such analysis should not overlook the effect of longer distance walking, as amputees are encouraged to walk for a longer distance [3,4]. 4) Finite element modeling requires the input of forces acting on the prosthesis to calculate the interface stress between the residual limb and the prosthetic socket. Such models should consider that the force data are subjected to changes after some walking.

Some results of this study indicated notable changes but did not reach statistically significant differences. Small sample size was one reason, and future studies should involve more amputee subjects. It should also be noted that five out of six subjects used solid-ankle cushioned heel (SACH) feet, which are one of the cheapest and most commonly used feet. It is interesting to observe if energy storage and return (ESAR) feet would produce different results. The subjects walked on the treadmill during the long-distance walking. This allowed monitoring for safety and the experiments to be performed in a smaller space. It has been suggested that treadmill walking might lead to a gait pattern different from that of over-ground walking, because the treadmill belt might force the subject to walk at a faster cadence and shorter step length [28]. However, we did not observe any significant change in the spatial-temporal gait parameters in the subsequent over-ground walking test. Our subjects walked at self-selected walking speed without using the handrails during the 30 minutes of walking. Previous research had found that such conditions would not diminish the intrinsic stride dynamics as compared to natural over-ground gait [29]. The undesirable adaptation effect of treadmill walking in gait pattern was not supported.

## 5. Conclusion

This study investigated how trans-tibial amputee gait changed after long-distance walking. Reductions in the sound side ankle rocker progression and push-off were observed and these would be attributable to plantar-flexors fatigue. This was compensated by a greater effort on the prosthetic side. There was significant increase in prosthetic side knee joint flexion and moment during loading response, which facilitated the anterior rotation of the prosthetic shank. The hip extensors at the prosthetic side provided more power to facilitate the propulsion. These maintained the spatial-temporal gait parameters, including walking speed, step length, stance time and cadence, unaffected after long-distance walking. To facilitate long-distance walking in trans-tibial amputees, it is recommended to introduce training of the sound side plantar-flexors endurance, and to improve the prosthetic design in order to assist anterior rotation of the prosthetic shank during loading response.

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